Year: 2020

A Review of the Applications of Additive Manufacturing Technologies Used to Fabricate Metals in Implant Dentistry

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Abstract: PURPOSE To review the primary additive manufacturing (AM) technologies used to fabricate metals in implant dentistry and compare them to conventional casting and subtractive methods. METHODS The literature on metal AM technologies was reviewed, and the AM procedures and their current applications in implant dentistry were collated and described. Collection of published articles about metal AM in dental field data sources: MEDLINE, EMBASE, EBSCO, and Web of Science searched. All studies related to AM technology description, analysis, and evaluation of applications in implant dentistry, including AM titanium (Ti) dental implants, customized Ti mesh for bone grafting techniques, cobalt-chromium (Co-Cr) frameworks for implant impression procedures, and Co-Cr and Ti frameworks for dental implant-supported prostheses were reviewed. RESULTS Literature has demonstrated the potential of AM technologies to fabricate dental implants, root-analog implants, and functionally graded implants; as well as the ability to fabricate customized meshes for bone grafting procedures. Metal AM technologies provide a reliable method to manufacture frameworks for implant impression procedures. Co-Cr and Ti AM frameworks for implant-supported prostheses provide a clinically acceptable discrepancy at the implant-prostheses interface. CONCLUSIONS Additional clinical studies are required to assess the long-term clinical performance, biological and mechanical complications, and prosthetic restoration capabilities of additively manufactured dental implants. Moreover, further studies are needed to evaluate their long-term success and survival rates and biological and mechanical complications of AM implant-supported prostheses.

DOI: https://doi.org/10.1111/jopr.13212

Originally published at:
Revilla-León, Marta; Sadeghpour, Mehrad; Özcan, Mutlu (2020). A Review of the Applications of Additive Manufacturing Technologies Used to Fabricate Metals in Implant Dentistry. Journal of Prosthodontics, 29(7):579-593.
DOI: https://doi.org/10.1111/jopr.13212

A review of the applications of additive manufacturing technologies used to process metals in implant dentistry

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Conflict of interest

The authors did not have any conflict interest, financial or personal, in any of the materials described in this study.

Funding

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

Keywords

3D printing; Additive manufacturing technologies; Electron Beam Melting; Implant dentistry; Selective Laser Melting

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Abstract

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Methods. The literature on metal AM technologies was reviewed, and the AM procedures and their current applications in implant dentistry were collated and described. Collection of published articles about metal AM in dental field data sources: MEDLINE, EMBASE, EBSCO, and Web of Science searched. All studies related to AM technology description, analysis, and evaluation of applications in implant dentistry, including AM titanium (Ti) dental implants, customized Ti mesh for bone grafting techniques, cobalt-chromium (Co-Cr) frameworks for implant impression procedures, and Co-Cr and Ti frameworks for dental implant-supported prostheses were reviewed.

Results. Literature has demonstrated the potential of AM technologies to fabricate dental implants, root-analog implants, and functionally graded implants; as well as the ability to fabricate customized meshes for bone grafting procedures. Metal AM technologies provide a reliable method to manufacture splinted frameworks for implant impression procedures. Co-Cr and Ti frameworks for implant-supported prostheses provide a clinically acceptable discrepancy at the implant-prostheses interface.

Conclusions. Additional clinical studies are required to assess the long-term clinical performance, biological and mechanical complications, and prosthetic restoration capabilities of additively manufactured metal frameworks and dental implants. Moreover, further studies are needed to evaluate their long-term success and survival rates and biological and mechanical complications of AM implant-supported prostheses.

Additive manufacturing (AM) technologies offer an alternative to conventional casting and subtractive fabrication methods.[1-4] AM technologies provide several advantages when compared with conventional processes, including freeform fabrication capabilities, minimized material waste, lightweight design, and elimination of production steps.[5] However, the technology also has disadvantages including limited building platform size, slow build rates, and disputed dimensional accuracy. There is also significant labor required for application design, setting process parameters, and necessary post-processing procedures.[5,6]

The International Organization for Standardization (ISO/TC261) defines AM as a "process of joining materials to make objects from three dimensional (3D) model data, usually layer upon layer, as opposed to subtractive manufacturing methodologies".[7-9] This article reviews the foremost AM technologies used to manufacture metal in the dental field as well as their applications in implant dentistry, such as fabricating implants, frameworks for implant impressions procedures, and frameworks for tooth- and implant-supported prostheses.

Five different databases were selected to perform the search of articles, namely MEDLINE, EMBASE, EBSCO, and Web of Science. The following MeSH terms, search terms, and their combinations were used in the search: "[MeSH] dental prostheses" OR "dental" OR "dentistry" AND "[MeSH] cobalt chromium" OR "[MeSH] titanium" OR "metal additive manufacturing" OR "3D metal printing" OR "metal printing" OR "direct metal laser sintering" OR "selective laser melting" OR "[MeSH] printing, three-dimensional" OR "electron beam melting". A total of 906 tittles and abstracts were reviewed; 123 articles were included in the present review.

All studies related to AM technology description, analysis, and evaluation of applications in implant dentistry were reviewed, including AM titanium (Ti) dental implants, customized subperiosteal Ti implants, customized Ti mesh for bone grafting techniques, cobalt-chromium (Co-

Cr) frameworks for implant impression procedures, and Co-Cr and Ti frameworks for dental implant-supported prostheses.

Metal AM technologies

Powder bed fusion (PBF) technology is the most common AM category used to process Co-Cr and Ti alloy metals in implant dentistry and includes selective laser sintering (SLS), selective laser melting (SLM), and electron beam melting (EBM) procedures.[1,9]

SLS technology involves a high-powered laser (Na:YAG laser) beam that selectively melts metal powder into a thin and solid layer (20-100 μm). Then, another layer of metal powder is deposited, and the next slice of the metal object is fused with the first. This procedure repeats until the metal framework is manufactured.[10-13] The manufacturing temperature does not reach the melting point of the processed metal, therefore, the metal powder sinters instead of fully melting.[14]

SLM technology employs high-quality lasers (CO_2 or Nd:YAG laser) that allow the complete melting of the metal powder in an inert chamber with purified argon or nitrogen.[13-18] The build platform is warmed to a temperature usually up to 200 °C.[19,20] The particle size of the powder ranges between 20 to 60 μm.[19] The SLM process accumulates high internal stresses due to manufacturing thermal changes and a post-processing heat treatment has been recommended.[21]

EBM technology employs a focused electron beam to melt layers of powder in an inert vacuum chamber with purified argon. During fabrication, an elevated temperature of about 700 ºC is sustained in the printer chamber to minimize temperature gradients and reduce residual stresses. First, a tungsten filament is heated to over 3000 ºC, triggering electrons to be emitted and accelerated due to the potential difference between an anode and cathode. The electrons are focused using magnetic coils to form a narrow high energy beam that strikes the surface of the metal powder. At this point, the kinetic energy transferred through friction creates the necessary heat to melt the metal powder.[15,18] The metal powder grain size typically used ranges from 45 to 150 μm.[22-24]

AM objects present a characteristic surface roughness resulting from the fabrication process which can vary depending on the technology employed. During SLM and EBM procedures, the surrounding metal powder that is not melted sinters to the surface of the AM object (Fig 1A). Therefore, the metal surface also contains partially melted powders.[23] Post-processing procedures for the purpose of surface modification have been recommended, namely mechanical abrasion, electro-polishing, sandblasting (Fig 1B), computer numerical control (CNC) machining (Fig 1C), chemical mechanical polishing, and laser polishing.[23] To the authors' best knowledge, sandblasting procedures are most commonly used when post-processing tooth-supported frameworks, while CNC machining is most commonly used to shape the implant interface when developing frameworks for implant-supported prostheses (Figs 2AB).[3]

There are numerous variables which are specific to each PBF device[18] including printing parameters such as laser beam absorption and reflections coefficients, energy source and power, melting temperature, thermal conductivity, temperature reached, printer chamber characteristics, metal powder particle morphology, layer thickness, and printing orientation.[18-30] Particle morphology influences metal powder performance, including flowability and packing efficiency, which consequently influence final component properties.[20,31-33] Another variable that should also be considered is the usage history of the metal powder. After the desired object is additively manufactured, the remaining non-melted powder can be recycled or discarded.[34] However, there is no specific protocol for the recycling procedures nor quality control. In addition, there is limited information available regarding how the grain characteristics of a recycled powder would impact the mechanical characteristics of the AM object.[34-38]

Conventional versus AM procedures

Subtractive and additive procedures have been previously described as hardware or software dependent, where performance is determined by the capability of a 3D CAD file to be produced as a 3D object. However, the conventional casting method is more dependent on the technician fabrication ability.[39] A few studies have compared the accuracy of metal AM in manufacturing with subtractive methods; although dental studies have reported similar manufacturing accuracy, none of the AM or subtractive procedures obtained a perfect match to the CAD file.[40,41]

The dental literature comparing microstructural, mechanical and electrochemical properties of casted, milled, and PFB AM metal dental alloys is scarce.[3,39,42-52] Moreover, the chemical composition of the different Co-Cr and Ti dental alloys vary due to the different manufacturing procedures complicating direct comparisons.[3,44] Different authors reported that the manufacturing method has a strong effect on the metal microstructure.[39,42-52] However, comparable data for the microstructure, thermomechanical history, and the extent of internal porosity for the three techniques was not available.[39]

Al Jabbari et al[44] evaluated the metallurgical and interfacial categorization of Co-Cr dental alloys fabricated using casting, milling, or AM SLM procedures. Radiographic evaluations revealed no internal porosities on the milled and AM specimens and gross porosities on the casted specimens. Microstructures were dependent on the manufacturing technique employed. Together with the γ phase matrix, a second phase, believed to be the Co_3Mo phase, was also observed by SEM and subsequent XRD analysis. Cr_7C_3 and $Cr_{23}C_6$ carbides were also identified via XRD analysis in the casted and milled specimens. Furthermore, significant differences were found among hardness values obtained from the groups tested where the Vickers hardness mean value for the casted group was 320 ±12 HV, for the milled group was 297 ±5 HV, and for the AM SLM group was 371 ±10 HV.

Takaichi et al[46] analyzed the microstructure and mechanical characteristics of casted and SLM AM Co-Cr dental alloys. Significant differences were found between the casted and the AM groups. Furthermore, build orientation and laser energy also demonstrated significant effect on the microstructure and mechanical properties of the AM specimens.

Kim et al[48] evaluated and compared the microstructural characteristics and mechanical properties of the Co-Cr metal alloys obtained by 4 different manufacturing procedures, namely casting, milling, SLM AM, and milled/sintered procedures. Chemical composition, microstructure, and mechanical properties were dependent on the manufacturing technique. Furthermore, SLM samples obtained the highest mean ultimate tensile strength follow by the milled/sintered, casted, and milled groups. The highest mean yield strength was found in the SLM AM group (580 ±50 MPa), followed by casted (540 ±20 MPa), milled/sintered (510 ±20 MPa) and milled (480 ±20 MPa) specimens. The percent elongation values were higher in the SLM AM group (32 ±2%) and by milled/sintered (27 ± 2%) groups compared with the casted (10 ±2%) and milled (2.3 ±0.7%) groups. The milled/sintered specimens obtained the highest mean Young's modulus (270 ±30 GPa), followed by the casted (260 ±20 GPa), milled (230 ±40 GPa), and SLM AM (200 ±10 GPa) specimens.

Han et al[49] evaluated the mechanical properties of casted, milled, and SLM AM Co-Cr dental alloys. Results showed that SLM AM specimens obtained the highest elongation fracture (13 ±1%) and Vickers hardness (399 ±24 Hv 10) values compared to casted and milled specimens. However, the milled group obtained the highest Young´s modulus (253 ±14 GPa).

Bilgin et al[50] measured the fracture resistance of casted, milled, and SLM AM Co-Cr post-cores. The metal post and cores were adhesively cemented on four extracted mandibular human teeth. Milled specimens obtained the highest fracture resistance mean values (959.26 ±110.79 N), followed by SLM AM (689.4 ±57.44 N), and casted groups (608.89 ±51.8 N).

Choi et al[51] evaluated the microstructural and mechanical properties of casted, milled, and SLM AM Co-Cr alloys. Cast samples revealed the highest Vickers hardness (455.88 ±37.08 Hv) followed by AM (413.10 ±8.77 Hv) and milled (243.4 ±8.97 Hv) groups. The mean ultimate tensile strength was highest for the milled specimens 1442.39 ±13.25 MPa, and the AM group presented 1411.12 ±17 MPa although the cast group showed the lowest value 831.51 ±41.10 MPa. Furthermore in 0.2% yield strength, the cast group exhibited from 770.18 MPa to 897.39 MPa, the AM group showed from 1384.74 MPa to 1438.64 MPa, and the milled group from 1427.86 MPa to 1459.22 MPa, which was the highest value. The cast and the AM groups exhibited 0.59 mm and 0.87 mm elongation, respectively, which were lower than total elongation. Lastly, the elastic modulus was highest for the AM group (67.0 GPa) followed by the milled group (61.0 GPa), and the cast group (59.0 GPa).

Zhou et al[52] also assessed the mechanical properties and microstructures of casted, milled, and SLM AM Co-Cr dental alloys. This in-vitro study also confirmed that the microstructures and mechanical properties of the Co-Cr alloys were dependent on the fabricating methods. In contrast with the casted and milled samples, the AM group displayed improved mechanical properties and microstructure. The AM group obtained the highest 0.2% yield strength (790 ±11 MPa) followed by casted (520 ±30 MPa) and milled (495 ±20 MPa) groups. Also, AM group obtained the highest ultimate tensile strength (1072 ±18 MPa) followed by casted (658 ±44 MPa) and milled (638 ±25 MPa) groups. Furthermore, the AM group showed the highest elongation percentage (12.7

±1.9%) followed by milled (11.1 ±1.0%) and casted (8.0 ±0.4%) groups; and the highest microhardness mean values (475.3 ±10.2 Hv 10) followed by milled (325.2 ±17.8 Hv 10) and casted (323.7 ±27.2 Hv 10) groups.

Compared with the conventional casting procedures, PBF AM technologies offer several advantages, such as a higher metal density, prevention of casting defects, decreased fabrication time and expenses, and minimization of human errors.[39,42-52] Studies which have evaluated Co-Cr and Ti metals fabricated through these AM technologies concluded that their mechanical properties were better than those using conventional casting techniques.[39,42-52] However, limited dental literature compared the properties of the metal manufactured using conventional, subtractive, and additive procedures.

From the economic standpoint, non-dental studies have determined that additional manufacturing costs using AM technologies compared to conventional methods include material, labor cost, machine expense, and energy consumption.[5,6] Material costs represent the major cost in metal AM procedures, with labor cost at 2-3% and energy consumption at less than 1%.[6] High production costs in AM technologies are due to slow build rate and the high cost of metal powder.[5,6] There is presently no normalized data regarding manufacturing costs within the dental field where, except for dental implants, most parts are completely customized to each patient.

Application of AM technologies in implant dentistry

In implant dentistry, AM is being evaluated to manufacture non-customized dental implants (similar to current commercial implants) as well as patient-customized metal devices. Research and industry concerns lie in determining where AM can substitute or generate new manufacturing systems.[53-55] AM technologies have been selected to manufacture Ti implants,[55-81] customized subperiosteal Ti implants,[82-86] customized AM Ti meshes for bone grafting techniques,[87-95] Co-Co frameworks for implant impression procedures,[96,97] and Co-Cr and Ti implant frameworks for implant-supported prostheses.[98-102]

With the introduction of AM technologies, different studies have analyzed the potential to manufacture Ti AM dental implants (Table 1),[55-73] customized designs that replicate the tooth's root shape (Table 2),[74-81] and customized subperiosteal Ti implants[82-86] (Table 3). Furthermore, the mechanical properties, osteoconduction, and bone augmentation properties of titanium porous lattice structures have been evaluated.[102]

Different in-vitro and animal studies,[17,53,54,56,60-62,64,72] as well as clinical studies[57,60,65-71] have analyzed the utility of AM titanium implants. Authors concluded that AM technologies are an alternative for manufacturing custom implants, providing adequate and controlled porosity levels, superficial roughness that promotes new bone formation, and improves the osseointegration process. Several clinical studies have described a high incidence of prosthodontic complications while using AM titanium implant procedures (Table 1).[67,68,70] Current challenges in additively manufactured dental implants include surface characteristics, dimensional accuracy, and high manufacturing cost.[54] Further studies are needed to define a standardized protocol to fabricate dental implants using AM technologies. Furthermore, authors concluded the benefits of the AM technologies for these purposes were not yet clearly described.[56]

Lin et al[59] evaluated the mechanical properties of an AM Ti graded solid implant design, with a control porosity distribution along the body of the implant. The implant design involved two main areas with a dense exterior layer and a partially sintered layer with different volume and porosity distribution at the interior. The concept design involved a combination of hardness and strength to withstand implant insertion forces and a more flexible interior structure to support the stiffness of the implant. The implant design attempted to provide similar mechanical properties of the cortical and trabecular bone.

Furthermore, in-vitro and animal studies,[73,74,76,77] as well as clinical studies[75,79-81] have analyzed the utility of AM root-analog titanium implants (Table 2). Similar to the design of commercial implants, AM root-analog implants provide a promising alternative from an osseointegration standpoint. However, from a prosthetic standpoint, only cemented restorations are possible due to the one-piece implant design.

A clinical study assessed the peri-implant soft tissues around the AM titanium implants retrieved from 12 patients.[68] Authors found collagen fibers oriented perpendicularly to a distance of 100 μm from the surface, where they became parallel, running in several directions. In some portions, only a few collagen fiber bundles appeared to be oriented perpendicularly or obliquely to the plane of the section. Under scanning electron microscopy, an intimate contact of the fibrous matrix with the implant surface was evident, and some collagen bundles could be seen to bind directly to the metal surface.[68] Furthermore, the porous surface of the AM implants seemed to be conducive to the direct attachment of the collagen fibers on the surface.

Several studies clinically evaluated AM subperiosteal implants (Table 3).[82-86] Cerea and Dolcini[83] performed a retrospective clinical study on the survival and complication rates of customized subperiosteal AM Ti implants in 70 patients with 2 years of follow-up. Authors reported 98% implant survival rate, 1.4% biological complications rate, and an 8.9% prosthetic complication rate. Mangano et al[84] described an AM customized subperiosteal Ti implant for the fabrication of an implant-supported fixed dental prostheses (FDP). Authors reported a case series of 15 partially edentulous patients that received the customized subperiosteal implant in the posterior mandible and a cemented milled zirconia implant-supported FDP. Within the 1-year follow-up, authors reported a 100% implant survival rate and a 30% prosthetic complications rate.

Non-resorbable prefabricated Ti barriers have been described for guided bone regeneration (GBR) techniques.[104,105] However, the shaping of prefabricated Ti mesh for a GBR technique can be difficult and time consuming, and it has been reported that soft tissues, such as mucous membranes, could also be easily injured.[106] With the incorporation of AM technologies, AM Ti customized mesh has been documented for bone grafting procedures (Table 4).[87-95]

Otawa et al[89] evaluated the dimensional accuracy of customized AM Ti mesh. The thickness of the polished device was 0.3 mm with a mesh aperture of 1.0 mm. Results obtained demonstrated that x- and y-axes reproduction presented higher accuracy compared to the z-axis. The mean accuracy value errors reported was 139 μm.

Sumida et al[90] fabricated and compared the clinical performance of customized AM with conventional Ti meshes for bone augmentation procedures in 26 patients. The customized AM meshes were fabricated with a thickness of 0.5 mm and with 1.0 mm diameter pores; the screw holes for positioning the AM mesh with fixation screws had a 1.5 mm diameter; and it was manually polished until 0.3 mm thickness was achieved. During implant placement, the Ti meshes were positioned into the implant site along with autologous bone. Authors reported a consistently higher surgery time for conventional Ti mesh compared to customized AM mesh surgeries due to manual bending of the conventional mesh. Furthermore, conventional mesh procedures were associated with 15.4% higher incidence of mucosal dehiscence with infection compared to AM procedures, however this difference was not statistically significant.

Inoue et al[93] evaluated the feasibility of the customized Ti mesh sheets for alveolar bone reconstruction in two patients. On the first clinical procedure, a customized SLM Ti mesh sheet was positioned at the same time of placement of a commercial implant. Six months after surgery, a CBCT analysis demonstrated adequate bone morphology under the mesh, and it was decided to be left in the mouth. On the second clinical procedure, a customized SLM Ti mesh was used for bone augmentation procedures. Four months after the surgery, the AM Ti device was removed, and three commercial implants were placed. An adequate bone morphology was observed on the postoperative CBCT.

The application of AM technologies offers several advantages to conventional techniques, namely predictable digital planning and design of the customized mesh, accurate additive manufacturing, shorter surgical times, and easy adaptation of the customized mesh.[3,87-95] Nonetheless, further clinical studies are desirable to measure the feasibility, success, and complication rates of customized Ti mesh for bone augmentation procedures.

Recent publications described a method for a complete-arch implant impression technique where the Co-Cr metal framework was manufactured using AM technology (Fig 3A).[95,96] The application of AM technology provides several advantages compared to conventional implant impression techniques, namely simple digital design of the framework, an accurate manufacturing procedure to reproduce the digital design of the splining framework, the ability to control impression material thicknesses, and easy placement in the patient´s mouth with a controlled splinting procedure to the impression abutments.[96]

AM technologies can be also selected to fabricate Co-Cr and Ti implant-supported prostheses.[3,96-101] However, because of the surface roughness of the AM metal frameworks, the success of these technologies relies on their combination with the subtractive technologies.[3,96-102] The combination of the additively manufactured metal framework and the posterior milling of the implant interface have allowed the application of AM technologies in implant dental prostheses (Fig 3BC).

Revilla-León et al[99] analyzed the distortion at the implant abutment-prosthesis interface of Ti frameworks fabricated with SLM and EBM processes using a coordinate measuring machine (CMM). The mean implant abutment-prosthesis discrepancies were 67 ±13.5 μm for the SLM technology and 60.2 ±18.5 μm for the EBM technology. Within the limitations of the study, the authors concluded that the Ti frameworks for a complete-arch implant-supported prosthesis manufactured using either SLM or EBM procedures demonstrated clinically acceptable implant abutment-prosthesis discrepancies. In a later study, Revilla-León et al[101] analyzed the misfit of complete-arch Co-Cr frameworks fabricated using three SLM technologies. The implant abutment-prosthodontic interface discrepancy was evaluated with a CMM machine. The SLM technologies evaluated obtained a mean 3D implant abutment-prosthodontic interface discrepancy ranging between 47.26 to 73.77 μm, which could be considered clinically acceptable.

Ciocca et al[102] evaluated the trueness and precision of Co-Cr 6-implant-supported frameworks fabricated using subtractive and additive procedures. The mean 3D positioning errors ranged from 8 to 22 μm for the SLM AM technique and from 20 to 35 μm for milling procedures.

An interesting feature of AM metal framework is the capability to manufacture a surface texture on the AM metal. Alsheghri et al[107] described a Y-shaped interlocking feature to increase the mechanical retention between the AM metal and the acrylic resin material. This study reported an increase in strength of the Co-Cr/PMMA interface from 2.3 MPa (flat interface) to 34.4 ±1 MPa, which constituted 85% of the tensile failure strength of PMMA (40.2 ± 1 MPa).[107]

Further studies are needed to evaluate the long-term success and survival rates and biological and mechanical complications of AM metal frameworks used in implant-supported prostheses.

Conclusion

Metal AM technologies have potential for various dental implant applications, such as metal frameworks for implant impression procedures and dental prostheses. AM technologies can also be used to fabricate both commercial-style and customized dental implants. Limited information is available regarding the applications for AM technologies for the fabrication of dental implant prostheses, as well as customized titanium meshes for bone augmentation procedures. Furthermore, future studies are desirable to assess their accuracy, reproducibility, clinical outcome over time.

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TABLES

Table 1. Literature documentation for AM titanium dental implants studies published.

Author	AM Technology	Study Design	Implant Type	Number Specimens	Results
Traini et al[17]	SLM (EOS)	In vitro	NA	10 disks, 20 bars (compact and porous bars)	• The mean porosity: 28.7 ±2.2% of the metal surface. • Young's modulus on porous bars: 77±3.5GPa and on compact bars 104±7.7GPa.
Chen et al[53]	SLM (SLM 125HL, SLM Solutions GmbH)	Part 1: FEA Part 2: Animal study	SLM solid 3 porous SLM implants (3 different porosity design)	Part 2: 16 rabbits	FEA: • Static analysis: Stress distribution pattern comparable among the groups (von Misses stress between 121-141 MPa). Micromotion was higher on porous implants than solid implants. • Dynamic analysis: Porous implant presented higher implant fatigue than solid implants. Animal study: Porous implants effectively increase the implant-bone contact area. Osseointegrated surface area was higher on porous implants than solid implants. Irregular reticular cancellous bone and newly formed bone grew inside some of the pores and hollow structures

Wang et al[60]	SLM (SLM Solutions)	Part-1: FEA and in vitro Part-2: Animal study	SLM implant with 2 designs: porous and solid with surface porous	Part 1: FEA Part 2: Rabbits	Part 1: FEA and in-vitro • Von Mises stress distribution was similar in both SLM solid (216 MPA) and porous (276 MPa) implant designs. • Elastic modulus porous (96 MPa) smaller than solid (111 MPa) implant design. Part 2: Animal study • Bone was present into the surrounding implant pores and the interconnected pores and exhibited a reliable osseointegration.
Hyzy et al[61]	DMLS (EOS)	Part-1: In vitro Part-2: Animal study	DMLS implant Milled implant Commercial implant (AB Dental)	Part 1: Discs, DMLS and commercial implants Part 2: Rabbits	Part 1 • DMLS implant presented higher surface roughness and wetability. • Pull out forces presented no significant differences between the 2 implant designs. Part 2 • Osseointegrations was achieved for both implant groups. Significant higher BIC values, osteoblast differentiation, and maturation on DMLS implants.
Shaoki et al[62]	SLM (SLM125HL; SLM solutions GmbH)	Part-1: In vitro Part-2: Animal study	SLM implant Milled implant Commercial implant (Nobel Biocare Speedy Replace RP)	Part 1: Discs Part 2: 13 beagle dogs 72 implants (n=24)	Part 1 • Superficial roughness (Ra) (SLM): 10.65 ±2.3 µm • Water contact angle (SLM): 86.97 ±2.65º • Good cell proliferation (osteoblast activity): Osteoblast proliferation and alkaline phosphatase activity were significantly enhanced on the SLM surface. Part 2 • BIC ratio SLM: 35.98%. No significant difference among groups.

					Removal torque SLM (45.41 Ncm) was significantly lower that of than Nobel-S implants and higher than that of milled implants.
Fukuda et al[72]	DMLS (EOSINT M270; EOS)	Part-1: In vitro Part-2: Animal study	Cylindrical channel DMLS implants	10 cylindrical channel DMLS implants, 8 beagle dogs	Part 1 • Newly formed bone around the DMLS implants • Relationship between osteoinduction and interconnection size in the range 500-1200 µm Part 2 • New bone formation was observed in all the channels of all treated DMLS implants. The highest observed osteoinduction occurring at 5 mm from the end of the implants.
Cohen et al[64]	DMLS (EOS)	Animal study	Solid implant Porous implant	20 implants (n=10)	Superficial roughness (Ra): 2.66 µm (solid implants) and 2.47 µm (porous implants). Porosity (porous implants): 68.6 ±0.8% Implants were osseointegrated. Porous specimens presented significantly higher bone volume formation than solid specimens.
Shibli et al[66]	DMLS (SLM) (metal printer not provided)	In vivo (DMLS implants were retrieved with a trephine after 8 weeks of placement)	DMLS implant with (for immediate loading) or without a solid abutment	12 completely edentulous patients with 2 DMLS implants per participant	• DMLS implants showed healthy surrounding bone. The newly formed bone showed early stages of maturing and remodeling. Osteoblast were connected to the newly bone formed. • BIC ratio ranged from 18.09% (submerged DMLS implant) to 54.5% (immediate loaded DMLS implant)

Mangano et al[57]	DMLS (SLM) (TixOs; Leader-Novaxa)	In vivo (DMLS implants were retrieved with a trephine after 8 weeks of placement)	DMLS implant	1 patient with 2 DMLS implants	Histologic examination revealed osteoblasts were observed close to the newly formed bone. The implant surface showed superficial debris or particle inclusions in the surrounding tissue close to the bone area. BIC ratio of 62.20%.
Mangano et al[65]	DMLS (SLM) (TixOs; Leader-Novaxa)	In vivo (DMLS implants were retrieved with a trephine after 8 weeks of placement)	DMLS one-piece implant	12 completely edentulous patients, 24 DMLS implants	• Collagen fibers, in the form of bundles, were oriented perpendicularly up to a distance of 100 µm from the surface where they became parallel, running in several directions. • Collagen fibers appeared to form a dense chaotic three- dimensional network running in different, more or less parallel directions to the surface. Under scanning electron microscopy, an intimate contact of the fibrous matrix with the implant surface was evident, and some collagen bundles could be seen to bind directly to the metal surface. • The surface of the implant was characterized by a disordered succession of irregular, rounded protrusions, narrow crevices, and intercommunicating pores.
Mangano et al[67]	DMLS (SLM) (TixOs; Leader-Novaxa)	Prospective clinical study (3-year follow-up)	DMLS implant	35 patients with a completely edentulous arch, 120 DMLS implants	Implant survival rate: 97.4% Biological complications rate: 7.1% Prosthodontic complication rate: 17.8% After 3-year follow-up, the mean distance between the implant shoulder and the first visible bone-to-implant contact was 0.62 ±0.28 mm.

Mangano et al[68]	DMLS (SLM) (TixOs; Leader-Novaxa),	Prospective clinical study (1-year follow-up)	DMLS implant one-piece implant (ball attachment)	24 patients with a completely edentulous mandible, 96 DMLS implants	Implant survival rate: 98.8% Implant success rate: 97.8% Prosthodontic complication rate: 58.3%
Mangano et al[69]	DMLS (SLM) (TixOs; Leader-Novaxa),	Prospective clinical study	DMLS implant one-piece implant (ball attachment)	62 patients, with a completely edentulous mandible, 231 DMLS implants	Mean follow-up period: 2.7 years after loading Implant survival rate: 96.9% The mean distance between the implant shoulder and the first visible bone-to-implant contact was 0.38 – 0.25 and 0.62 – 0.20 mm at the 1- and 4-year follow-up examinations, respectively. Biologic complications rate: 6.0% Prosthodontic complication rate: 12.9%
Mangano et al[70]	DMLS (SLM) (Eosint270, EOS)	Prospective clinical study (5-years of function, removed due to implant fracture)	DMLS Implant	2 DMLS implants	Compact, mature lamellar bone was observed over the majority of the specimens. BIC% of 66.1 ±4.5%.
Mangano et al[71]	DMLS (SLM) (TixOs; Leader-Novaxa)	Prospective clinical study (1-year follow-up)	DMLS implant	68 patients, 201 DMLS implants	Prosthetic rehabilitations of 201 DMLS implants included: 105 crowns, 45 FDPs, and 2 fixed complete-arch prostheses. Implant survival rate: 99.5%

NA, Not applicable; BIC ratio, Bone implant contact ratio; FEA, Finite element analysis; FDP, Fixed dental prostheses

Table 2. Literature documentation for AM titanium customized root analog or multi-rooted analog implants studies published.

AUTHOR	AM TECHNOLOGY	STUDY DESIGN	IMPLANT TYPE	NUMBER SPECIMENS	RESULTS
Chen et al[73]	SLM (SLM125HL; SLM solutions GmbH)	Part-1: In vitro Part-2: Finite element analysis	Root-analog (RA) Root-analog threaded (RTA)	20 implants (n=10)	Part-1 results • Relative density: >99% • Superficial roughness (Ra): 4.74 ±0.01 µm • Bend strength: 1067 ±42 MPa • Dimensional accuracy (diameter/length): 60 ±23 µm and 210 ±75 µm. • Implant stability quotient and pull-out strength was higher with a RTA design Part-2 results: Better stress distribution and lower maximum micromotions were observed for the RTA model than for the root-analog implant model.
Ramakrishnaiah et al[74]	EBM (A2; Arcam)	In vitro	Root-analog	NA	Micro porosities characteristics of the specimens stimulated bone ingrowth and improved bone bonding. The surface chemical composition presented low levels of carbon and high wetting capabilities.
Moin et al[76]	SLM (LayerWise; 3D Systems)	In vitro	Root-analog	1 specimen	0.27% surface increased of the 3D surface model compared to the original tooth. Towards the more apical regions of the root the divergence gradually increased up to more than -0.15 mm.

Peng et al[77]	SLM (AM250, Renishaw)	Animal study	Multi-rooted analog (MRI) Commercial implants	33 rabbits	MRI specimens presented higher bone volume density compared to commercial specimens. MRI specimens obtained a peak value of 48.41 %. MRI specimens showed denser surrounding bone growth compared to commercial specimens; after 4 and 8 weeks, bone tissue had grown into the pore structures and root bifurcation areas. MRI specimens obtained a push-out forces from 294.7 to 446.5 N and maximum mean torque forces from 81.15 to 289.57 N, while the commercial specimens obtained 34.79 to 87.8 N respectively.
Figliuzzi et al[75]	DLMF (Leader; Novaxa)	Case report (1-year follow-up)	Root-analog	1 patient	The specimen was clinically stable with no infection, and radiographically unchanged peri-implant marginal bone level and no peri-implant radiolucency was observed.
Moin at al[79]	SLM (Replica Implant System)	Nonrandomized, non- controlled prospective pilot case series	One-piece root-analog with zirconia abutment	6 patients 6 root-analog implants	1 implant/patient lost; data not included No intraoperative complications reported. 1 implant presented mobility with peri-implant infection after 5 weeks of placement. 4 remaining implants were successful after 1 year after loading. Bone loss after implant placement 0.59 mm; after provisional delivery -0.36 mm; after 1-year function -0.31 mm.
Mangano et al[80]	SLM (EOS)	Case report	One-piece root-analog implant	1 patient	Immediate root-analog implant placement (position first maxillary right premolar) with an interim cemented crown. After 3 months, a cemented implant-supported metal-ceramic crown was delivered.

Mangano et al[81]	DMLS (Leader implants)	Prospective clinical study (1-year follow-up)	One-piece root-analog	15 patients, 15 root-analog implants	Implant survival rate: 100% The mean distance between the implant shoulder and the first visible bone-to-implant contact was 0.7± 0.2 mm Prosthodontic complication rate: 0%

NA: Not available

Table 3. Literature documentation for AM titanium customized subperiosteal implants.

AUTHOR	AM TECHNOLOGY	STUDY DESIGN	IMPLANT TYPE	NUMBER SPECIMENS	RESULTS
Gellrich et al[82]	AM Ti	Technique description	Subperiosteal implant	NA	Digital design description: First, implants are positioned based on the prosthodontic planning; secondly, a wire-frame framework is designed and adapted on the patient´s recipient bone.
Cerea et al[83]	DMLS (ProX-DMP100; 3D systems)	Retrospective clinical study	DMLS subperiosteal Ti implant	70 patients, 70 DMLS implants	Follow-up period: 2 years Implant survival rate: 95.8% Post-operative complications rate: 5.7% Biological complications rate: 1.4% Prosthetic complications rate: 8.9%
Mangano et al[84]	DMLS (ProX-DMP100; 3D systems)	Prospective clinical study	DMLS subperiosteal Ti implant	10 patients, 10 DMLS implants, 10 cemented FDPs	Follow-up period: 1 year Implant survival rate: 100% Post-operative complications rate: 10% Prosthetic complications rate: 20%
Mounir et al[85]	EBM (Arcam AB)	Prospective clinical study	DMLS subperiosteal Ti and PEEK implant	10 patients (n=5), Ti and PEEK implants	Follow-up period: 1 year Implant survival rate: 100%
Mommaerts et al[86]	SLM (CADskills)	Case report	SLM subperiosteal Ti implant	1 patient	Complete maxillary subperiosteal Ti implant AM complete-arch implant-supported interim restoration

FDP, Fixed dental prostheses; PEEK, Polyether ether ketone

Table 4. Literature documentation for custom AM titanium (Ti6AlV4) meshes for bone augmentation procedures.

AUTHOR	AM TECHNOLOGY	STUDY DESIGN	MESH DESIGN	NUMBER SPECIMENS	RESULTS
Otawa et al[89]	SLM (Printer not provided)	In vitro study	Custom AM Ti mesh	10 specimens	Goal: Evaluate the feasibility of using SLM to produce Ti customized devices for bone grafting Results: • 292 µm: Maximum error obtained • 139 µm: Average accuracy error measured.
Ciocca et al[87]	SLM (Eosint M270; EOS)	Case report	Custom AM Ti mesh	1 patient	The mean post-op height value was 14.18 mm; the mean post-op buccal–palatal width value was 10.37 mm. The mean value difference between the pre- and post-op heights was 2.57 mm, and the mean value difference between the pre- and post-op widths was 3.41 mm
Sumida et al[90]	SLM (Eosint M270; EOS)	Prospective study	Custom AM Ti mesh Conventional Ti mesh	26 patients (n=13)	Higher surgery time for conventional Ti mesh compared to customized AM mesh surgeries. Success rate of AM mesh: 92.3% Success rate of conventional mesh: 76.9%
Sagheb et al[91]	Not provided	Retrospective clinical study	Custom AM Ti mesh	17 patients	Mean vertical augmentation of 6.5 ±1.7 mm and a mean horizontal augmentation of 5.5 ±1.9 mm. 33% patients presented an exposure of the TM after a period ranging from 5 to 12 weeks from first-stage surgery

Inoue K al[93]	SLM (Printer not provided)	Case report series	Custom AM Ti mesh	2 patient	Successful bone grafting procedures were reported and analyzed in a post-operative CBCT. • Case 1: Customized Ti mesh sheet was placed at the same time of implant placement. Ti mesh was left in the patient´s mouth. No complication. CBCT confirmed bone morphology under mesh. • Case 2: Customized Ti mesh sheet for bone augmentation. After 3 months, it was removed and 3 implants were placed. Bone regeneration was observed on the buccal site.
Hartmann et al[95]	SLM (Yxoss CBR®l ReOss)	Retrospective clinical study	Custom AM Ti mesh	55 patients	25% patients presented with exposure of the mesh Precise fit, shorter time of surgery, predictable outcome and good acceptance of the surgical procedure

FIGURES

Figure 1. Environmental scanning electron microscope (FEI QUANTA 200; Thermofisher Scientific) images. A, As-built titanium (Ti-6Al-4V) specimen fabricated using EBM technology (Arcam Q10Plus; Arcam) visualizing the partially melted powders on the surface of the AM metal. B, SLM AM (EOS M 290; EOS) Co-Cr (EOS CoCr SP2 Powder; EOS) specimen after sandblasting procedures. C, Titanium milled specimen.

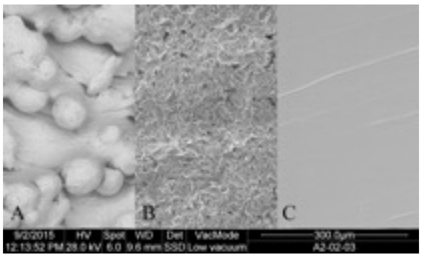

Figure 2. Metal frameworks for implant supported prostheses. A, SLM AM Co-Cr metal framework as printed. B, SLM AM CoCr metal framework after sandblasting and milling the implant interface.

A B

Figure 3. Metal 3D printing applications in implant dentistry. A, 3D printed metal splinting structure for a complete arch implant impression technique. B, Implant-supported Co-Cr SLM AM metal framework for a metal-ceramic fixed dental prosthesis. C, Complete-arch implant-supported titanium EBM AM framework for a metal-resin or metal-composite resin fixed dental prosthesis.

A B C